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Effect of hip and knee position on tensor fasciae latae elongation during stretching: An ultrasonic shear wave elastography study

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Abstract

Background: Decreased flexibility of the tensor fasciae latae is one factor that causes iliotibial band syndrome. Stretching has been used to improve flexibility or tightness of the muscle. However, no studies have investigated the effective stretching position for the tensor fasciae latae using an index to quantify muscle elongation in vivo. The aim of this study was to investigate the effects of hip rotation and knee angle on tensor fasciae latae elongation during stretching in vivo using ultrasonic shear wave elastography.

Methods: Twenty healthy men participated in this study. The shear elastic modulus of the tensor fasciae latae was calculated using ultrasonic shear wave elastography. Stretching was performed at maximal hip adduction and maximal hip extension in 12 different positions with three hip rotation conditions (neutral, internal, and external rotations) and four knee angles (0°, 45°, 90°, and 135°).

Finding: Two-way analysis of variance showed a significant main effect for knee angle, but not for hip rotation. The post-hoc test for knee angle indicated that the shear elastic modulus at 90° and 135° were significantly greater than those at 0° and 45°.

Interpretation: Our results suggest that adding hip rotation to the stretching position with hip adduction and extension may have less effect on tensor fasciae latae elongation, and that stretching at >90° of knee flexion may effectively elongate the tensor fasciae latae.

Highlights

- An effective stretching position for the tensor fasciae latae was examined.
- Shear wave elastography was used as an index to quantify muscle elongation
- Hip adduction, extension and >90° of knee flexion is the most effective position.
- Shear wave elastography is an effective method to investigate muscle elongation.
- High reliability was confirmed for the measurement of shear wave elastography.

Keywords

Ultrasonic shear wave elastography

Tensor fasciae latae

Stretching

1 Introduction

Iliotibial band (ITB) syndrome, one of the most common overuse injuries of the tensor fasciae latae (TFL) and ITB, often causes pain within the lateral portion of the knee joint. Competitive runners and cyclists (Ellis et al., 2007) as well as patients with knee osteoarthritis (Vasilevska et al., 2009) are reportedly at high risk of developing ITB syndrome. In a recent review of the mechanism of ITB syndrome, an abnormal increase in the compression forces between the ITB and the lateral epicondyle causes irritation and inflammation in the tissue beneath the ITB (Louw and Deary, 2014). Furthermore, two studies that investigated hip biomechanics using computer modeling reported that ITB hardness was influenced by the tension of the TFL (Birnbaum et al., 2004 and Fetto et al., 2002). Thus, it is important that the tightness and flexibility of the TFL are kept normal to avoid ITB hardness.

Static stretching is a common method for improving muscle flexibility (Nakamura et al., 2014). The stretching position and maneuver should be determined based on kinesiology and anatomy of the muscle. Because the function of the TFL is hip flexion, abduction, and internal rotation (Paré et al., 1981), stretching for TFL involves the joint motion opposite to the function of the muscle, i.e. hip extension, adduction, and external rotation. The two commonly used stretching positions for the TFL are; hip adduction, hip extension, and 90° knee flexion (i.e., Ober test), and hip adduction, hip extension, and full knee extension (i.e., modified Ober test). The two commonly used stretching positions for the TFL are (1) hip adduction, hip extension, and 90° knee flexion, which is consistent with the position used in the Ober test, and (2) hip adduction, hip extension, and full knee extension, which is consistent with the position used in the modified Ober test. The Ober test and modified Ober test are orthopedic examinations to check the shortening of the ITB length (Ober, 1936 and Kendall et al., 1970). However, no studies have quantitatively investigated the effective stretching position for the TFL in vivo because it was conventionally impossible to directly and noninvasively assess the quantified elongation of the hip muscles, which have multidirectional joint movements. The traditional method estimating muscle elongation such as range of motion or passive torque may be influenced by many factors such as other muscles, ligaments, and the joint capsule crossing the joint. Hence, elongation of individual muscles could not be assessed using these traditional measurements (Koo et al., 2014).

The recent development of a new ultrasound-based technique, namely ultrasonic shear wave elastography, allows the non-invasive and reliable measurement of muscle elasticity (Bercoff et al., 2004). Previous studies verified a strong linear relationship between passive muscle elongation measured using traditional methods and the shear elastic modulus measured by ultrasonic shear wave elastography in vitro (Eby et al., 2013 and Koo et al., 2013) or in vivo (Maïsetti et al., 2012 and Koo et al., 2014). Furthermore, some studies have investigated the effect of stretching on the muscle, using ultrasonic shear wave elastography (Taniguchi et al., 2015; Akagi and Takahashi, 2014 and Akagi and Takahashi, 2013). Therefore, ultrasonic shear wave elastography is a valid technology for investigating changes in muscle elongation in vivo.

A previous study investigating the effective stretching position reported that muscle elongation was influenced by the muscle moment arm (Umegaki et al., 2014) as well as the kinesiology and anatomy of the muscle. Additionally, the TFL has been reported to have the moment arms of hip internal rotation (Mansour and Pereira, 1987) and knee extension (Spoor and van Leeuwen, 1992). Therefore, we hypothesized that the TFL could be further stretched by adding hip external rotation and knee flexion to hip adduction and hip extension. The objective of this study was to investigate the effects of hip rotation and knee angle on the shear elastic modulus of the TFL during stretching using ultrasonic shear wave elastography in vivo.

2 Methods

2.1 Subjects

Twenty healthy men participated in this study [mean age, 23.3 (1.6) years; mean height, 172.9 (4.4) cm; mean weight, 66.6(6.2) kg]. Subjects were non-athletes and had not performed any excessive exercise. Subjects with a history of orthopedic or nervous system disease in their limbs were excluded. All subjects provided written informed consent. This study protocol was approved by the ethics committee of Kyoto University Graduate School and the Faculty of Medicine (E-1162).

We calculated the sample size needed for two-way analysis of variance (ANOVA) with repeated measures (effect size = 0.25, α error = 0.05, power = 0.8) using G* power 3.1 software (Heinrich Heine University, Duesseldorf, Germany). The results showed that 18 subjects were required; therefore, 20 subjects were recruited in this study to account for potential withdrawal.

2.2 Experimental protocol

All procedures were performed by the same two investigators: one performed the stretching maneuver, while the other measured the shear elastic modulus to ensure reproducibility.

Each subject lay in a supine position on a bed with the trunk securely fixed by a non-elastic band. The right lower limb was chosen for the measurement. The lower limb of the non-measurement side (the left side) was held at 125° of hip flexion and maximal knee flexion to maintain posterior pelvic tilting. The rest position (REST) was defined as that with the hip in a neutral position (i.e., 0° hip extension, 0° abduction, and neutral rotation) and the knee in full extension. For all of the stretching positions, the hip was kept in maximal adduction and maximal extension. Stretching was measured in the combinations of three hip rotation conditions (neutral rotation, maximal internal rotation, and maximal external rotation) and four knee angles (0°, 45°, 90°, and 135°) for a total of 12 different conditions (Figure 1).

Regarding the joint movement order during the stretching maneuver, the knee was flexed first, followed by maximal hip adduction, hip extension, and hip rotation. During the stretching maneuver, the hip joints were moved to the maximal angle at which the subjects felt no discomfort or pain. The knee angles were fixed during the stretching maneuver using a Donjoy knee brace (DJO Global Inc., Vista, CA), which is a knee brace with a dial lock to maintain each angle during the stretching maneuver for rehabilitation. Each stretch was performed in a random order to preclude the effect of the measurement sequence. Since a previous study reported that >2 minutes of stretching decreased muscle stiffness (Nakamura et al., 2013), each stretch was performed for <15 seconds to prevent effects of changes in muscle stiffness on the TFL.

2.3 Assessment of the shear elastic modulus

The shear elastic modulus of the right TFL was measured at REST and in each of the 12 stretching positions using ultrasonic shear wave elastography (Axiporer; SuperSonic Imagine, Axi-en-Provece, France) with an ultrasound transducer (50-mm-long SL-15-4 linear ultrasound transducer). The measurement site was defined as the midpoint between the anterior superior iliac spine and the greater trochanter of the femur. The region of interest (ROI) was set up near the central point of the muscle belly in the image. A10-mm-diameter circle was drawn around the center of the ROI. The mean shear wave propagation speed (m/s) within the circle was automatically calculated. The shear elastic modulus (G) was converted from the shear wave propagation speed (V) using the following equation:

$G = \rho V^2$

where ρ is the muscle mass density, which is presumed to be 1,000 kg/m^3 (Gennisson et al., 2005; Nordez et al., 2008 and Nakamura et al., 2014).The shear elastic modulus was measured twice and the mean value was used for the analysis. Previous studies reported that the shear elastic modulus calculated by shear wave elastography was strongly correlated with the degree of muscle elongation (Eby et al., 2013 and Koo et al., 2013).

2.4 Measurement reliability

Reliability of the shear elastic modulus measurements was ascertained using the intraclass correlation (1, 1) ($ICC_{1, 1}$). $ICC_{1, 1}$ was calculated by the shear elastic modulus of the two measurements at each of the REST and stretching positions.

2.5 Statistical analysis

Statistical analysis was performed using SPSS (version 18.0; SPSS Japan Inc., Tokyo, Japan). To determine whether TFL was elongated in each stretching position, differences in shear elastic modulus between the REST and each stretching position were assessed using the paired student’s *t*-test with Bonferroni revision. Two-way ANOVA with repeated measures using two factors (hip rotation [three positions] × knee angle [four positions]) was used to determine the effects of hip rotation and knee angle on the shear elastic modulus. When a significant main effect was found, the post-hoc test was performed. A confidence level of 0.05 was used in all of the statistical tests. For the shear elastic modulus in the stretching position, the effect size was calculated from the formula: $(X_1 – X_2) / \sqrt{\ } [(S_1^2 + S_2^2) / 2]$ using G* power 3.1 software (Heinrich Heine University, Duesseldorf, Germany). In the above formula, X_1 is mean of each stretching position, X_2 is REST, S_1 is standard deviation of each stretching position and S_2 is standard deviation of REST.

3 Results

3.1 Measurement reliability

The reliability of the shear elastic modulus for the REST and stretching positions is shown in Table 1. The $ICC_{1, 1}$ was 0.932–0.986 for all positions, which was significant.

3.2 Effect of hip and knee position on stretching – induced tensor fasciae latae elongation

The shear elastic modulus of each stretching position is shown in Table 2 as mean (standard deviation), with the effect size. The shear elastic modulus at REST was 13.4 (5.2) kPa. When the shear elastic modulus was compared between the REST and stretching positions, the shear elastic modulus of each stretching position was significantly higher than that of the REST position ($p < 0.05$). Two-way ANOVA showed a significant main effect of the knee angle ($F = 15.35$, $p < 0.01$) but not hip rotation ($F = 1.13$, $p = 0.33$), with no significant interaction between hip rotation and knee angle ($F = 0.87$, $p = 0.52$). The post-hoc test for knee angle indicated that the shear elastic modulus at 90° and 135° were significantly higher than those at 0° and 45°. However, there were no significant differences between 0° and 45° or between 90° and 135° (Figure 2).

4 Discussion

4.1 Measurement reliability

In this study, the $ICC_{1,1}$ of the measurement was 0.932–0.986 for all positions, which was significant. An ICC value of 0.40 is generally considered poor reliability, 0.40–0.75 is moderate to good, and 0.75 is excellent (Leong et al., 2013). We consider the data in this reliability study valid because the $ICC_{1,1}$ observed here was similar to that in a previous study (Leong et al., 2013).

4.2 Effect of hip and knee position on stretching – induced tensor fasciae latae elongation

This is the first study to examine the effective stretching position of TFL using the shear elastic modulus measured by ultrasonic shear wave elastography, which was defined as the degree of muscle elongation in vivo. The main findings of this study were that the stretching positions with hip adduction and extension may effectively elongate TFL, which could more effectively be stretched by the addition of >90° of knee flexion at this stretching position than by the addition of hip rotation.

We hypothesized that the TFL could be further stretched by adding hip external rotation and knee flexion to hip adduction and hip extension. However, our hypothesis was only partially proven because one part of the hypothesis stating that the TFL could be further stretched by adding hip external rotation was disproved and the another part stating that the TFL could be further stretched by adding knee flexion was confirmed. The moment arm can be calculated by dividing the amount of elongation of the muscle tendon unit (MTU) by the changes in joint angle (tendon excursion methods) (Maganaris et al., 2000). Therefore, the greater the moment arm and changes in joint angles are, the more elongated the MTU is (Umegaki et al., 2014). As for the moment arm of the TFL, the moment arms of hip abduction and hip flexion are large (Dostal et al., 1986), whereas the moment arm of hip internal rotation is small (Mansour and Pereira, 1987) or nonexistent (Dostal et al., 1986). Thus, due to moment arm of hip rotation, no effects of hip rotation on the TFL might have been observed in our study. On the other hand, our results indicated that the shear elastic modulus of the TFL was influenced by knee angle and that a stretching position with >90° of knee flexion may effectively elongate the TFL. The TFL has a moment arm on knee extension through the range of the knee joint motion (Spoor and van Leeuwen, 1992), a finding that is consistent with our result that indicated greater elongation of the TFL as the knee is flexed.

Our results agree with the findings of a previous study (Gajdosik et al., 2003) that indicated that the stretching position with 90° of knee flexion (i.e., Ober test) was more effective for stretching the TFL than that with a 0° knee angle (i.e., modified Ober test). However, Wang et al. (2006) reported that the stretching position with a 0° knee angle (i.e., modified Ober test) was more effective than that with 90° of knee flexion (i.e., Ober test), which is inconsistent with our result. This previous study (Wang et al., 2006) compared the properties of ITB at the stretching position between knee flexion and knee extension using an ultrasonographic image and concluded that the modified Ober test was more effective for stretching the TFL. In our study, the muscle belly of the TFL was compared to determine the effective stretching position, which might have caused this inconsistency. Furthermore, this inconsistency suggests that there might be a difference in tension applied between the TFL and ITB with knee angle changes during stretching. It is possible that the inconsistency is caused by the complicated structures of the ITB and surrounding tissues. The ITB has many attachments other than the TFL, including the gluteus maximus (Vieira et al., 2007), vastus lateralis, biceps femoris, lateral patellar retinaculum, patella, and patellar tendon (Baker et al., 2011). Therefore, further research is required to determine the difference in the stretching position between the TFL and ITB, with a focus on the hip and knee positions.

Our findings showing the effective stretching position of the TFL may be useful in the clinical and sports settings. However, this study had some limitations. First, the subjects in this study were healthy young men without a history of orthopedic or nervous system disease. Therefore, similar effects cannot always be expected in elderly people or patients with a limited range of motion. Second, we investigated only the acute effects of stretching position; therefore, it is unclear whether a long-term intervention program affects TFL elongation. Further research is required to determine the intervention effect in elderly people and patients with a limited range of motion.

5 Conclusion

Here we investigated the effects of hip rotation and knee flexion during stretching on muscle elongation of the TFL using the shear elastic modulus measured by ultrasonic shear wave elastography. Our findings suggest that adding >90° of knee flexion to the stretching position with hip adduction and extension may effectively elongate the TFL.

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Figure captions

Figure 1. Representative image of the stretching position. The subject is seen stretching to maximum hip adduction, extension, neutral rotation, and 45° of knee flexion in the supine position.

Figure 2. Comparison of the shear elastic modulus (kPa) in the post-hoc test for knee angle. Although no significant differences are noted between 0° and 45° (p = 0.13) and between 90° and 135° (p = 1.00), the shear elastic modulus are significantly higher at 90° and 135° than at 0° (p < 0.01) and 45° (p = 0.05).

*p < 0.05, **p < 0.01, significant difference among knee angles.

Stretching position	ICC	95% CI
REST	0.980	0.950-0.992
Hip N, Knee 0°	0.934	0.844-0.973
Hip N, Knee 45°	0.965	0.916-0.986
Hip N, Knee 90°	0.962	0.909-0.985
Hip N, Knee 135°	0.975	0.938-0.990
Hip IR, Knee 0°	0.986	0.967-0.995
Hip IR, Knee 45°	0.968	0.923-0.987
Hip IR, Knee 90°	0.934	0.844-0.973
Hip IR, Knee 135°	0.950	0.880-0.980
Hip ER, Knee 0°	0.980	0.950-0.992
Hip ER, Knee 45°	0.948	0.875-0.979
Hip ER, Knee 90°	0.963	0.911-0.985
Hip ER, Knee 135°	0.932	0.840-0.972

Table 1. Reliability of shear elastic modulus measurements.

ICC, intraclass correlation coefficient (1, 1); 95% CI, 95% confidence interval;

N, neutral rotation; IR, internal rotation; ER, external rotation

		Hip		
		Neutral rotation	Internal rotation	External rotation
Knee	0°	24.6 (8.0), effect size: 1.7	26.8 (15.7), effect size: 1.1	23.4 (9.2), effect size: 1.3
	45°	30.2 (10.5), effect size: 2.0	29.0 (11.1), effect size: 1.8	29.2 (11.1), effect size: 1.8
	90°	38.1 (11.0), effect size: 2.9	38.1 (14.5), effect size: 2.3	35.2 (12.1), effect size: 2.3
	135°	38.4 (17.5), effect size: 1.9	32.7 (12.9), effect size: 2.0	35.2 (12.3), effect size: 2.3

Table 2. Shear elastic modulus (kPa) of the tensor fasciae latae in the stretching position. Two-way analysis of variance showed a significant main effect of the knee angle but not hip rotation. Values are expressed as mean (standard deviation).

Figure1

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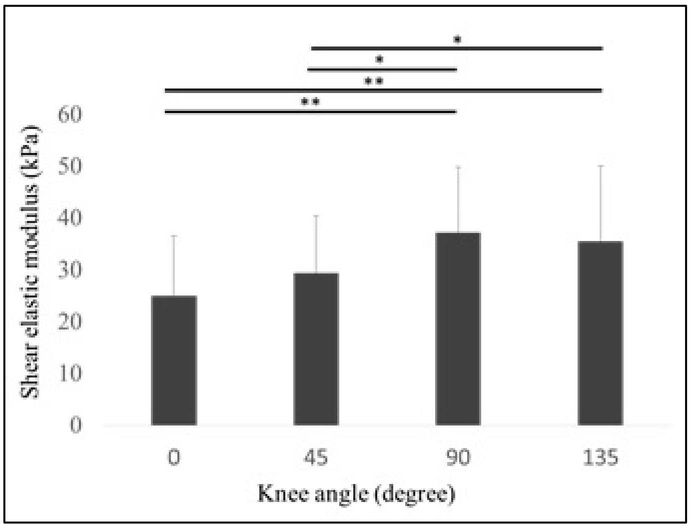

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